Comparison of 7+1ch tx arrays and Implementation for torso imaging at 3T - Four different coil geometries with phantom and human body

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Introduction

In previous studies, we proposed 8- and 16-channel tx body arrays for homogeneous excitation at 3T (1,2). After that, we enhanced the structure of the coil geometry to get a more homogeneous field with B1 shimming and by using a body size bottle phantom we also are able to use the advantage of B1 shimming over a large ROI. Here, we report the progress toward a case where 8 elements are placed in the space above the patient table as a transmit coil and while allowing space for a receive coil (or array) to maximize SNR from the chest.

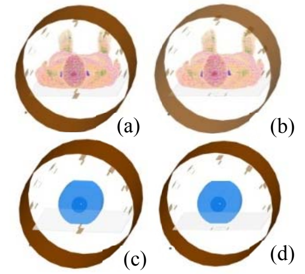


Figure 1: 4 different array structures and 2 different objects simulated in xFDTD program. (a) 7+1 strip line (coil A) with body mesh, (b) 7 strip line + 1 loop (coil B) with body mesh, (c) 7 +1 strip line, fitted array (coil C) with water bottle, and (d) 7 strip line, fitted array + 1 loop (coil D) with water bottle.

Methods

We modeled 4 different array structures and 2 different objects for field simulation at 3 T (123.3 MHz) with the finite difference time domain (FDTD) method (xFDTD; Remcom, Inc.). The whole mesh matrix size is 180x180x438 in the x, y, and z directions and each model consists of a model of an adult male (3) at 5x5x5 mm^3 resolution and RF shield with a 68cm diameter. The lengths of the inner conductor and ground plate for each element are 42cm and 50cm, respectively. Each element is driven with 2 current sources at opposite ends of the element and in opposite directions connecting the element to its ground plate. The distance between the ground plates of the opposing elements is 59cm. Figure1 shows 4 different models used in simulation. All array structures adapted 7 (above the patient) + 1 (under the patient) structure. Compared to the previously proposed structure (8 channels above the patient equi-azimuthal angle(1)), this structure can give higher signal intensity and receive sensitivity on the back (spine) parts. In Figure 1a and b the 7 upper coil elements are of the same strip line structure (Figure 2b), but to avoid the interference to the existing Siemens body coil, all elements are placed between the rods of the body coil. In Figure 1c and d, a more fitted array model is shown. The three coil elements on the top of the array are moved 1 inch toward object. This movement not only provides the enhanced isolation between existing body coil and tx array, but also gives advantages in coil sensitivity over the top part of the object. In Figure 1b and d the 25 x 10 cm rectangular loop coil element (Figure 2c) is used as an 8th channel transceiver coil. The water bottle used for coil-tuning and imaging is designed to avoid the dielectric resonance effects in imaging at 3T, as described in (1). For the phantom, we used 14.2 kg table sugar, 400 g NaCl, and 14.2 g $CuSO_4$ in 14.2 liters of distilled water. The measured relative permittivity and conductivity of the bottle was 63.86 and 0.4889 S/m, respectively. The properties of the phantom were used to model a digital phantom in xFDTD. The distances of long axis and short axis of the bottle cross-section were 10 inches and 9 inches, respectively. For B1+ shimming we used a homemade program in Matlab (The Mathworks). The whole mesh matrix size is 180x180x438 in the x, y, and z directions and each model consists of a model of an adult male (3) at 5x5x5 mm^3 resolution and RF shield with a 68cm diameter which is used for the clinical MR system. The simulated 8-channel above-bed array coil was constructed with copper plates, wood, and Derlin. The coil array was designed for a Siemens 3T whole body system (Magnetom 3T, Siemens Healthcare, Erlangen, Germany) equipped with a 8 x 8 kW peak RF power amplifier (Analogic, Peabody, MA), and a custom-built T/R switch and pre-amplifier box (Stark Contrast, Erlangen, Germany). The relative phase differences of transmit chains were measured and compensated and images and B1 maps (4) were acquired for the water phantom and body mesh.

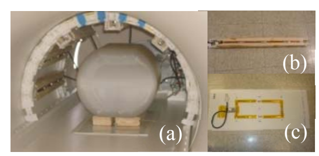
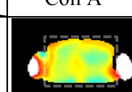
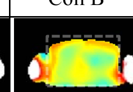
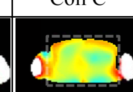
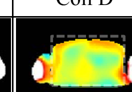
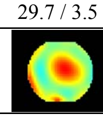
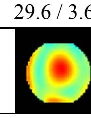
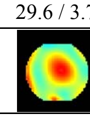
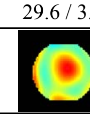


Figure 2: The picture of implemented tx array with water phantom and coil elements used. (a) the installed 7+1 tx array as Fig 1d. (b) strip line coil element. (c) rectangular loop coil element

Results and Discussion

Table 1 shows the resultant flip angle map after B1+ shimming over the body mesh and water phantom. Even the B1+ fields before shimming show quite different aspects to each other; the B1+ shimming methods we performed always provided a certain level of homogeneous field intensity over all the executed simulations. Compared to the body mesh model, the shimmed field with the water bottle model is more inhomogeneous. Although the phantom was designed to have permittivity notably less than water, the permittivity achieved is still likely above that of the chest on average, especially considering the low permittivity of lung. Regarding the issue of higher peak SAR over the area close to the coil elements (eg. the arm area), we could solve the problem with weighted B1 shimming. But in the point of efficiency of the coil element, it is a more proper way to replace the coil elements to the further the available position from the arm. We continue to search and test for a proper position and element for the torso and abdomen imaging applications.

Object \ Coil Structure		Coil A	Coil B	Coil C	Coil D
Body mesh	Flip angle map (After shimming)				
	Average/Std over the ROI	29.7 / 3.5	29.6 / 3.6	29.6 / 3.7	29.6 / 3.6
Water bottle	Flip angle map (After shimming)				
	Average/Std over whole area	29.5 / 5.6	29.5 / 5.5	29.5 / 5.9	29.7 / 5.7

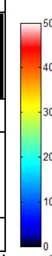


Table 1: The average and standard deviation of flip angle values before and after shimming of each array model with the water phantom and body mesh. B1+ shimming was done over the masked chest part (dashed square on the image). The targeted flip angle was 30 degrees.

Acknowledgements

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References

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